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Wearable inertial sensors to measure gait and posture characteristic differences in older adult fallers and non-fallers: A scoping review

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ABSTRACT

Background: Wearable inertial sensors have grown in popularity as a means of objectively assessing fall risk. This review aimed to identify gait and posture differences among older adult fallers and non-fallers which can be measured with the use of wearable inertial sensors. In addition to describing the number of sensors used to obtain measures, the concurrent anatomical locations, how these measures compare to current forms of clinical fall risk assessment tests and the setting of tests.

Methods: Following the development of a rigorous search strategy, MEDLINE, Web of Science, Cochrane, EMBASE, PEDro, and CINAHL were systematically searched for studies involving the use of wearable inertial sensors, to determine gait and postural based differences among fallers or those at high fall risk compared with non-fallers and low fall risk adults aged 60 years and older.

Results: Thirty five papers met the inclusion criteria. One hundred and forty nine gait and posture characteristic differences were identified using wearable inertial sensors. There were sensor derived measures which significantly and strongly correlated with traditional clinical tests. The use of a single wearable inertial sensor located at the lower posterior trunk, was most the most effective location and enough to ascertain multiple pertinent fall risk factors.

Conclusion: This review identified the capabilities of identifying fall risk factors among older adults with the use of wearable inertial sensors. The lightweight portable nature makes inertial sensors an effective tool to be implemented into clinical fall risk assessment and continuous unsupervised home monitoring, in addition to, outdoor testing.

1. Introduction

One in three adults over 65 years fall each year [1,2] and this increases to 40% of individuals over 80 years [3,4]. Falls are associated to a considerable increase in morbidity, immobility, mortality and loss of independence [5,6] and can also result in a heavy psychological burden due to a fear of falling [7]. Together these result in diminished mobility, social isolation and reduced quality of life [8,9] posing high economic burdens [10]. Therefore, primary prevention of falls is of paramount importance.

There are a variety of different clinical tests used to quantify fall risk. Examples of these include: the Timed-Up-and-Go test [11], Sit-To-Stand test [12], Dynamic Gait Index [13], Berg Balance Scale [14] and Tinetti Performance Oriented Mobility Assessment [15]. Despite being relatively quick and reliable whilst providing potentially relevant information on fall risk, clinical tests are steeped in a history of subjective observation [16]. Although, some tests such as the Timed-Up-and-Go test is measured by time taken and could be considered objective. Nevertheless, these clinical tests lack construct validity and fail to provide data from activities of daily living in a habitual environment setting, which will undoubtedly be different to a clinician's simulated assessment [17]. There have also been previous suggestions that clinical-based testing may be contaminated by the Hawthorne effect [18]. This is where participants perform differently and at times better than they normally would, due to awareness of being examined. Recent

Abbreviations: WIS, wearable inertial sensor; IMU, inertial measurement unit; Acc, accelerometer; RMS, root mean square; HR, harmonic ratio; AP, anteroposterior; ML, mediolateral; COM, centre of mass; SD, standard deviation; CoV, coefficient of variation; UAx, uniaxial accelerometer; DAx, dual axis accelerometer; Triax, triaxial accelerometer; L, lumbar; C, cervical; T, thoracic; S, sacrum; Vert, vertebrae; St, static; Dy, dynamic; NS, not stated; SS, self-selected; FP, fast pace; EO, eyes open; EC, eyes closed; ECF, eyes closed foam surface; NR, not reported; PD, Parkinson's disease; F, fallers; NF, non-fallers; FoF, fear of falling; GBVA, good binocular vision acuity; PBVA, poor binocular vision acuity

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research has suggested clinical assessments may not provide a true reflection of the individual being assessed. Giannouli et al. [17] identified that in older adults without any mobility impairments, mobility capacity related measures had little significance for predicting real-life performance. Additionally, Cofré Lizama et al. [19] reported the ability to perform accurate tracking of mediolateral centre of mass deteriorates with age, meaning that this potentially crucial factor may be missed with current clinical based assessments.

Recently, wireless and wearable technology, such as accelerometers and gyroscopes, have emerged as a potential alternative to clinical/laboratory testing. Such devices have the potential to be a worthy replacement for previously mentioned fall risk assessments and can even be used in conjunction with clinical tests; whilst providing objective data, resulting in more informed decisions regarding fall risk and subsequent fall preventative treatments. A wearable inertial sensor (WIS), is also known as a wearable inertial measurement unit (IMU). They are small, lightweight, inexpensive and does not require the arduous set-up times of traditional motion capture systems. Additionally, it possesses a long battery life and does not have to be confined to a clinic/laboratory. Therefore, WIS can provide continuous real time kinematic data during activities of daily living, as opposed to artificial movements of traditional fall risk assessment tests.

Current literature has identified the use of inertial sensors to detect when falls have occurred [9,20]. Despite potentially reducing time spent on the floor following a fall, the overall health consequences from falling will still remain. Therefore, primary prevention of falls is required and WIS-based technology may possess a pivotal role in achieving this. To successfully prevent falls from occurring whilst using WIS a thorough list of characteristic difference and potential fall risk factors, which have been measured using said WIS, is required. Recent reviews [8] and [21], provided insightful preliminary results regarding feature classification models and fall risk assessment and prediction. However, a rigorous search strategy was not provided, and systematic search results suggest that potentially meaningful papers may have been missed. Additionally, these reviews did not explore the potential clinical relevance of individual sensor derived features and how they correlate with traditional clinical assessments.

Therefore, the aim of this review was to identify, describe and evaluate a thorough list of characteristic differences and potentially fall risk factors related to gait and posture among older adults who are fallers and/or high fall risk, which have been measured with the use of WIS. In addition to, how they compare to current clinical assessments. The study also aimed to describe the number of WIS used, anatomical sensor locations and the setting of tests.

2. Methods

The review was conducted following the Preferred Reporting Items for Systematic Reviews and Meta-Analyses (PRISMA) guidelines. A scoping review was chosen due to the statistical heterogeneity of the studies within this field, deeming a standard systematic review with meta-analysis to be inappropriate. A protocol was developed following consultation with topic and methodological experts (https://ore.exeter.ac.uk/repository/handle/10871/26862).

2.1. Search strategy and eligibility criteria

The search strategy did not have data limits, ensuring a thorough search of the literature was performed. Table 1 highlights the search strategy developed and was applied in the following databases: MEDLINE, Web of Science, Cochrane, EMBASE, PEDro, and CINAHL.

Additionally, to ensure all relevant papers had been retrieved, a forward and backward search of the literature was also performed.

Studies were included within the review providing they successfully met the following selection criteria. All studies must include the use of WIS to identify gait and posture characteristics or fall risk factors, by comparing fallers and/or high fall risk older adults to non-fallers and low fall risk older adults aged over 60 years, who may be community-dwelling, in residential care or hospital patients. Additionally, if a study involved participants younger than 60 years they were still included, if the mean age minus one standard deviation was over 60 years [22]. No restrictions were placed on how measurements featuring WIS were obtained. This could be as participants performed a fall risk assessment task (clinical based) or any other activity away from the clinic/laboratory (e.g., continuous home walking or activities of daily living). All study designs were eligible for inclusion. No language restrictions were imposed on the inclusion criteria. Papers were excluded if they failed to meet any of the aforementioned inclusion criteria. Initial database searches were conducted on 7th January 2017, with an updated search performed on 20th April 2018.

Table 1
Example search strategy (OVID MEDLINE) – 24/12/2016.

	Accidental falls/ (48497)
1	Accidental falls/ (48497)
2	(Falls or faller$1 or fall risk factors).tw. (270649)
3	1 or 2 (295405)
4	Exp Aged/ (5572731)
5	(Senior$1 or elder* or old* or geriatric or geront*).tw. (6489416)
6	4 or 5 (10996364)
7	3 and 6 (153124)
8	Wearable.tw. (12766)
9	Inertial.tw. (18858)
10	Sensor*.tw. (1116595)
11	Track*.tw. (630433)
12	Device.tw. (854436)
13	Triaxial.tw. (5549)
14	Acceler*.tw. (870782)
15	Gyro*.tw. (13727)
16	Unit*.tw. (4373630)
17	8 or 9 or 10 or 11 or 12 or 13 or 14 or 15 or 16 (6996093)
18	Humans.sh. (18147175)
19	17 and 18 (965594)
20	7 and 19 (3085)

2.2. Study selection

Two reviewers independently screened all titles and abstracts of retrieved articles from the electronic database searches. Both reviewers then repeated the process whilst screening the full texts of the remaining papers. Any discrepancies were resolved via discussion when necessary. Data were extracted from included studies by one reviewer and checked by a second. The data management process was performed using EndNote X8.0.2 (Clarivate Analytics, USA).

2.3. Data extraction

Data were extracted on population characteristics (gender, age, body mass, and clinical condition when applicable), study characteristics (sample size, study duration and data collection setting, study design [prospective or retrospective]), methods used to measure characteristic differences or fall risk factors (quantity and type of inertial sensors used and their anatomical location(s)) and the type of activity performed when measurements were taken (clinical assessment; static or dynamic, or activities of daily living). A characteristic difference was extracted when a statistically significant difference between fallers/high fall risk versus non-fallers/low fall risk older adults were identified ($P < 0.05$). How these variables compared to current clinical assessments were extracted from studies who conducted correlation analysis between sensor derived and clinical assessment measures.

2.4. Data synthesis

Following data extraction a single reviewer independently synthesised data. All data were initially tabulated and subsequently reported descriptively [23] to answer the key questions of the review.

2.5. Risk of bias assessment

In order to determine the quality of the studies within the review a modified version of the Downs and Black Quality Index was used [24]. The original index was modified as it also evaluates clinical trials, these items were subsequently not included. The modified index was comprised of twenty questions related to information reporting (1–9), external validity (10 and 11), internal validity (12–15) and selection bias (16–20). Questions were answered with '1′ (yes) and '0′ (no). A score of 0–6 suggested a high risk of bias, 6–13 equals a moderate risk of bias and 14–20 proposed a low risk of bias.

3. Results

Database searches returned 7252 citations following the removal of duplications. One hundred and sixty six papers underwent full text scrutiny against the inclusion/exclusion criteria. Following this, 35 papers remained and were included within the review (Fig. 1).

The 35 included papers reported 23 studies featuring a comparison of fallers to non-fallers and twelve where subjects were classified as 'high fall risk' compared with 'low fall risk'. There were 2685 participants included although this excluded one study [25] where there was a lack of clarity over sample size, with sample sizes ranging from 12 to 260. Six studies failed to provide sufficient detail on test group characteristics [2,26–30]. 70 % (1823/2608) of participants were females, although gender was not reported in three studies [25,30,31].

Four different settings were used: a laboratory, home environment, local community centre and a hospital (Tables 2A and 2B). Two studies included mixed environments (laboratory and a home) [32,33]. Of the seven studies set in a home environment, two were short supervised sessions [28,34], whereas five studies involved continuous monitoring over two days to fourteen weeks [27,32,33,35,36].

Eight studies included static measures of fall risk [2,26,28,34,37–40], 30 utilised a form of dynamic movement [25,27–35,39,41–57] and three studies included both [28,34,39]. Studies typically involved clinical fall risk assessments, such as: Berg Balance Scale, Timed-Up-and-Go test, the Romberg test, walking a specified distance at a self-selected pace and/or quiet standing in different conditions (Tables 2A and 2B).

The number of inertial sensors used varied between studies, with a range of one to five sensors. A single triaxial accelerometer was most common, employed in fourteen studies [2,25,27,31,33,41,44,46,47,49–51,55,56] and two triaxial accelerometers were used in six studies [29,45,48,54,58,59]. Six studies used a single wearable IMU [35,38,39,52,53,57] and two studies used three [28,32] and five [34,37] IMUs each. Inertial sensors were predominantly placed on the posterior trunk [2,25,27,28,30–34,37–39,41,42,45–47,49,55,57–59]. Four studies failed to provide sufficient detail on trunk location, merely stating 'trunk' [37] and 'lower back' [25,33,55], respectively. See Tables 2A and 2B for a full listing of the number of WIS used, in addition to anatomical sensor locations.

Six studies included a prospective design of either six months [27,32–34] or twelve months, respectively [45,48] (Tables 2A and 2B).

There were 149 gait and posture differences identified from the 35 studies, as statistically significant differences were observed between groups whilst using WIS. Of this, 127 were dynamic measures (Table 3) whilst 22 were static measures (Table 4).

Identified characteristics and potential fall risk factors were categorised as temporal, spatial, linear acceleration, angular velocity, position and angle or energy variables. Some characteristics were statistically significant in multiple studies, but were counted only once within the aforementioned totals in the present review. Whereas, many other variables were identified (or measured) in only one study with some variation. Two of the 35 studies failed to reach statistical significance for any WIS based measures between fallers and non-fallers [46,48]. In the event of some studies witnessing a repetition of statistically significant results for the same variables in different conditions, such as: standing with eyes open or eyes closed and on a hard or foam surface [2,28,34,38], during different phases of a clinical assessment [49,52,57], different phases of turning courses [31] and also at different times during the day [28], they have each been recorded once. Tables 3 and 4, contain a full listing of all observed gait and posture differences measured using WIS and the corresponding studies they were identified in. Table 5 highlights the number of measures ascertained from WIS at identified anatomical locations. The lower trunk consisting of the third to fifth lumbar vertebrae, second lumbar vertebrae, sacrum and 'lower back' had the most identified sensor derived measures, with 102 of 172 measures (59.3%). The third to fifth lumbar

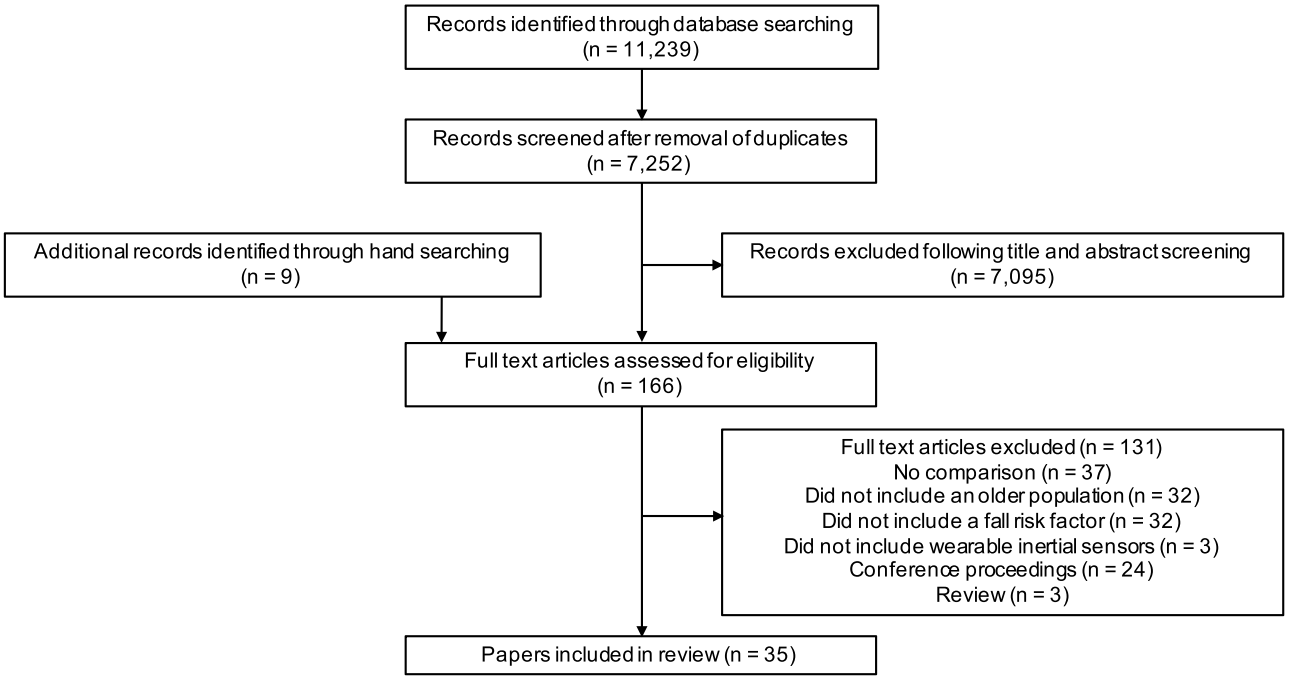

Fig. 1. Flow chart schematic illustrating the screening process.

Table 2A
Study characteristics including sample size, age, gender, height and body mass of included participants.

AUTHOR	SAMPLE	AGE (yrs)	GENDER	HEIGHT (cm)	WEIGHT (kg)
Fallers					
M/F					
Alqahtani et al. [26]	Total n = 29	87 (6)a	8/21	NR	NR
	Non-Fallers n = 10	NR	NR	NR	NR
	Fallers n = 19	NR	NR	NR	NR
Auvinet et al. [42]	Total n = 53	NR	20/33	NR	NR
	Healthy Older adults n = 33	77.2 (6.5)	18/15	163.1 (7.5)	63 (16.3)
	Fallers	80.7 (5.2)	2/18	156.5 (7.3)	59.8 (12.2)
Brodie et al. [43]	Total n = 96	NR	32/64	NR	NR
	Non-Fallers n = 61	80 (4)	22/39	162.5 (10)	67.6 (13.4)
	Fallers n = 35	79 (4)	10/25	159.8 (7.5)	66.8 (11.7)
Brodie et al. [44]	Total n = 18	NR	4/14	NR	
	Non-Fallers n = 11	84 (7.9)	4/7	162.6 (11.3)	71.6 (17.2)
	Fallers n = 7	82.2 (5.9)	0/7	154.8 (6.8)	71.3 (11.9)
Cho and Kamen, [40]	Total n = 16	NR	4/12	NR	NR
	Healthy Older n = 8	72.6	2/6	NR	NR
	Fallers n = 8	76.3	2/6	NR	NR
Cole et al. [58]	Total n = 30	NR	NR	NR	NR
	Older Adults n = 10	68.6 (2.2)	6/4	168.7 (2.7)	65.9 (3.1)
	PD Non-Fallers n = 10	66.5 (2.5)	6/4	168.5 (3.8)	67.9 (3.8)
	PD Fallers n = 10	69.3 (2.2)	6/4	165.7 (3.5)	65.9 (6.2)
Doheny et al. [28]	Total n = 40	71.4 (7.3)	17/19	NR	76.9 (11.3)
	Non-Fallers n = 21	NR	NR	NR	NR
	Fallers n = 19	NR	NR	NR	NR
Doi et al. [45]	Total n = 73	NR	16/57	NR	NR
	Non-Fallers n = 57	79.7 (8.2)	15/42	151 (10)	53 (9.7)
	Fallers n = 16	84.8 (5.9)	1/15	147 (10)	48.3 (9)
Gago et al. [37]	Total n = 36	NR	20/16	NR	NR
	Healthy n = 16	72.31 (7.08)	6/10	160 (11)	71.68 (9.07)
	Alzheimer's NF n = 9	73.56 (8.72)	7/2	152 (6)	68.9 (9.82)
	Alzheimer's F n = 11	77.64 (4.8)	7/4	153 (8)	65.01 (7.84)
Greene et al. [38]	Total n = 120	73.7 (5.8)	57/63	NR	NR
	Non-Fallers n = 55	73.27 (5.77)	NR	167.2 (9.05)	76.03 (13.78)
	Fallers n = 65	74.64 (4.8)	NR	165.65 (9.45)	77.2 (15.23)
Kojima et al. [25]	NR	NR	NR	NR	NR
Laessoe et al. [46]	Total n = 94	73.7 (2.9)	24/70	NR	NR
	Non-Fallers n =80	73.8 (2.9)	NR	NR	NR
	Fallers n = 14	73 (2.9)	NR	NR	NR
Latt et al. [29]	Total n = 99	NR	45/54	NR	NR
	Control n = 33	67 (4)	15/18	168 (3)	70 (4)
	PD Non-Fallers n = 33	63 (4)	15/18	170 (3)	73 (5)
	PD Fallers n = 33	67 (2)	15/18	169 (3)	68 (5)
Liu et al. [39]	Total n = 12	NR	5/7	NR	NR
	Healthy Young n = 4	21.75 (0.96)	1/3	167 (9)	64.07 (13.9)
	Healthy Older n = 4	73.25 (7.09)	2/2	171 (10)	71.89 (23.14)
	Fall Prone Older n = 4	74.5 (2.65)	2/2	173 (14)	73.71 (12.49)
Mancini et al. [32]	Total n = 35	NR	12/23	NR	NR
	Non-Fallers n =16	83.9 (7)	3/13	NR	NR
	One-time Fallers n = 12	86 (7)	4/8	NR	NR
	Recurrent Fallers n = 7	88.4 (8.8)	5/2	NR	NR
Matsumoto et al. [47]	Total n = 85	80 (7.3)	7/78	148.9 (7.3)	48.5 (8.9)
	Non-Fallers n = 51	79 (7.6)	4/47	149.6 (6.7)	49.3 (8.9)
	Fallers n =34	81.6 (6.4)	3/31	147.6 (6.8)	47.3 (8.1)
Mohler et al. [34]	Total n = 119	NR	24/95	NR	NR
	Non Frail NF n =20	74.7 (6.7)	4/19	NR	NR
	Non-Frail F n = 23	74.4 (6.6)	3/17	NR	NR
	Pre-Frail NF = 38	79.7 (8.5)	10/28	NR	NR
	Pre-Frail F n =19	79.4 (8.8)	4/15	NR	NR
	Frail NF n =10	86.6 (5.9)	3/7	NR	NR
	Frail F n = 9	80.9 (9.8)	0/9	NR	NR
O'Sullivan et al. [2]	Total n = 17	77 (7.5)	8/9	NR	NR
	Non-Fallers n = 12	NR	NR	NR	NR
	Fallers n = 5	NR	NR	NR	NR
Paterson et al. [48]	Total n = 97	68.73 (7.07)	0/97	161.07 (6.29)	69.78 (16.01)
	Non-Fallers n = 43	68.4 (7.31)	0/43	160.83 (5.23)	69.12 (18.47)
	Fallers n =54	69 (6.93)	0/54	161.27 (7.06)	70.31 (13.89)
Pozaic et al. [35]	Total n = 136	72.5 (5.6)	56/80	169.3 (9.1)	73.9 (14.7)
	Non- Fallers n = 123	72.4 (5.6)	52/71	169.7 (9.2)	74 (14.9)
	Fallers n = 13	74.2 (5.3)	4/9	165.8 (7.4)	72.8 (12.3)
Van Schooten et al. [27]	Total n = 169	75.4 (6.8)	81/88	NR	NR
	Retrospective				
	Non-Fallers n = 109	NR	NR	NR	NR
	Fallers n = 60	NR	NR	NR	NR

(*continued on next page*)

Table 2A (*continued*)

AUTHOR	SAMPLE	AGE (yrs)	GENDER	HEIGHT (cm)	WEIGHT (kg)
	Prospective				
	Non-Fallers n = 110	NR	NR	NR	NR
	Fallers n = 59	NR	NR	NR	NR
Weiss et al. [49]	Total n = 41	NR	NR	NR	NR
	Healthy Control	68.3 (9.1)	8/10	164 (7)	70.18 (6.4)
	Idiopathic Fallers	76 (3.9)	6/17	162 (6)	69.6 (10.9)
Weiss et al. [33]	Total n = 71	NR	25/46	NR	NR
	Non-Fallers n = 39	78.77 (4.39)	14/25	164 (6)	72.02 (13.36)
	Fallers n = 32	77.86 (5.09)	11/21	161 (9)	71.94 (12.29)
High Risk					
Asai et al. [59]	Total n = 260	71.9 (3.9)	114/146	156.2 (8.4)	57.4 (9.8)
	Non-FoF n = 202	71.6 (3.8)	101/101	157.2 (8.5)	58.1 (9.9)
	FoF n= 58	72.7 (3.9)	13/45	152.9 (7.5)	55 (8.9)
Bautmans et al. [50]	Total n = 121	NR	61/60	NR	NR
	Young n = 40	21.6 (1.4)	20/20	175.3 (8)	68.2 (17.1)
	Older n = 41	79.1 (4.9)	21/20	164.6 (7.7)	75.3 (9.3)
	Older Fall Risk n = 40	80.6 (5.4)	20/20	161.8 (12.1)	72.2 (15.2)
Brodie et al. [51]	Total n = 30	NR	18/12	NR	NR
	Young n = 10	29.6 (6.6)	6/4	171 (9.5)	68.2 (17.1)
	Healthy Older n =10	65.6 (6.9)	5/5	171 (9.7)	75.3 (9.3)
	Older Parkinson's n =10	67.1 (4.1)	7/3	166.3 (12.5)	72.2 (15.2)
Ganea et al. [52]	Total n = 106	NR	37/69	NR	NR
	Healthy Older n = 27	73 (5.03)	11/16	167 (9)	71.5 (13.5)
	Frail Older n = 79	80 (7.1)	26/53	162 (51)	67.2 (15.4)
Ishigaki et al. [53]	Total n = 95	75.4	9/86	149.1 (8.03)	52.36 (9.98)
	Stable n = 40	69.2 (5.7)	NR	150.87 (7.33)	53.89 (8.76)
	Unstable n =55	79.9 (7.1)	NR	147.59 (8.43)	51.22 (10.85)
Matsumoto et al. [41]	Total n = 223	73.6 (8.3)	82/141	154 (9.1)	53.1 (9.6)
	No LS = 182	72.2 (7.9)	71/111	156 (8.8)	53.7 (9.6)
	LS n = 41	79.5 (7.2)	11/30	148 (7.8)	50 (8.7)
Menz et al. [54]	Total n = 100	79.9 (4)	32/68	NR	NR
	Low risk n = 34	78.94 (3.44)	NR	NR	NR
	Moderate Risk n = 33	79.15 (3.83)	NR	NR	NR
	High Risk n = 33	81.48 (4.45)	NR	NR	NR
Moe-Nilssen and	Total n = 65	NR	26/39	NR	NR
Helbostad, [55]	Fit n = 33	73.1 (3.3)	20/13	170 (8)	NR
	Frail n = 32	80.5 (4)	6/26	164 (8)	NR
Senden et al. [56]	Total n = 100	NR	44/56	NR	
	Low Fall Risk n = 50	74.2 (5.1)	27/23	168 (9)	72.3 (12.7)
	High Fall Risk n = 50	78.9 (6.2)	17/33	167 (11)	70.3 (13.5)
Shin et al. [31]	Total n = 22	NR	NR	NR	NR
	GBVA n = 11	75.55 (7.27)	NR	149.11 (3.86)	50.35 (6.79)
	PBVA n = 11	77 (5.62)	NR	149.51 (4.21)	51.15 (5.22)
Yack and Berger, [30]	Total n = 39	NR	NR	NR	NR
	Young n = 19	24 (2.6)	NR	NR	NR
	Older n = 20	NR	NR	NR	NR
	Older Stable n = NR	78 (7)	NR	NR	NR
	Older Unstable = NR	77 (9.9)	NR	NR	NR
	Older Non-Fallers n = 5	NR	NR	NR	NR
	Older Fallers n = 5	NR	NR	NR	NR
Zakaria et al. [57]	Total n = 38	NR	20/18	NR	NR
	Low Fall Risk n = 17	63 (8.9)	NR	NR	NR
	High Fall Risk n = 21	71.1 (5.8)	NR	NR	NR

Data presented as mean (SD); NR = Not reported; PD = Parkinson's Disease; LS – Locomotive syndrome; GBVA = Good binocular vision acuity; PBVA = Poor binocular vision acuity; F = Fallers; NF = Non-fallers; FoF = Fear of falling.

vertebrae was the location that yielded the most measures across all studies (46/172, 26.7%). The second lumbar vertebrae had 26 measures, but this all came from a single paper [57]. The sternum and sacrum were also desirable locations for various parameters, respectively (20/172 11.6%). If multiple studies had the same location for one measure this was recorded once. However, it should be noted that some measures were determined using multiple anatomical locations, for example, walking speed was measured with a sensor located at the' lower back' [25,33], third to fifth lumbar vertebrae [27,29,31,42,45,49,59], second lumbar vertebrae [57], tenth thoracic vertebrae [58], sacrum [43,50,56] and head [51]. The study by Mohler et al. [34] was omitted from this analysis as five separate inertial sensors were simultaneously used for each measure they reported.

Five studies presented correlation analysis on WIS-based measures to traditional clinical assessment with mixed results [2,25,38,43,56]. Maximum entropy, mediolateral harmonic ratio and the eight step harmonic ratio variation were weakly correlated to walking speed. The root mean square (RMS) of the acceleration signal during quiet standing on a mat with eyes open yielded strong statistically significant correlations with the Timed-Up-and-go test but inversely, albeit strongly correlated with the Berg Balance Scale [2]. The RMS of the angular velocity signal and RMS of the acceleration signal were also strongly correlated with the Berg Balance Scale [38]. Whilst walking speed and step length were found to have strong positive correlations to the Tinetti scale [56].

Four studies did assess the test-retest reliability of WIS-based measures, finding moderate and high reliability [35,41,43,50].

The average score from the risk of bias assessment was 13.2 out of

Table 2B
Setting, tests performed, duration of test, number of inertial sensors and corresponding anatomical locations.

AUTHOR	SETTING(S)	TEST(S)	DURATION	SENSOR(S)	LOCATION(S)
Fallers					
Alqahtani et al. [26]	Laboratory	4 Standing balance tests Short physical performance battery	30 s – N/A – NS 10 s – 4 m – NS	1 DAx Acc	Iliac crest
Auvinet et al. [42]	Laboratory	Walking self-selected pace	N/A – 40 m – NS	2 UAx Acc (AP & ML)	L3 – L5 Vert
Brodie et al. [43]	Laboratory	Walking self-selected pace	N/A – 20 m – 2	1 Acc (NS)	Sacrum
Brodie et al. [44]	Home	Daily life walking	N/A – Continuous – 14 weeks	1 TriAx Acc	Sternum (skin)
Cho and Kamen, [40]	Laboratory	4 Standing sensory conditions	20 s – N/A – 5	2 UAx Acc (AP)	Sacrum (S2), forehead
Cole et al. [58]	Laboratory	Walking at 3 paces	N/A – 60 s – 1	2 TriAx Acc	T10, posterior head
Doheny et al. [28]	Home (supervised)	4 Standing tests Walking self-selected pace	30 s – N/A – 1 N/A – 3 m – 4	3 IMUs	L3 Vert (St), Left and right shin (Dy)
Doi et al. [45]	Laboratory *	Walking self-selected pace	N/A – 15 m – NS	2 TriAx Acc	C7, L3 vert
Gago et al. [37]	Laboratory	6 Standing Romberg conditions	30 s – N/A – NS	5 IMUs	Trunk, left and right thighs and shanks
Greene et al. [38]	Laboratory	2 Standing balance tests	40 s (EO) 30 s (EC) – N/A – 3 (EO) 4 (EC)	1 IMU	L3 Vert
Kojima et al. [25]	Laboratory	Walking self-selected pace, fast pace	N/A – 11 m – 3 (SS), 2 (FP)	1 TriAx Acc	Lower back
Laessoe et al. [46]	Community centre	Walking – 3 different speeds	N/A – 14 – 2	1 TriAx Acc	L3 Vert
Latt et al. [29]	Laboratory	Walking self-selected pace	N/A – 20 m – 2	2 TriAx Acc	Head (helmet), level of sacrum
Liu et al. [39]	Laboratory	3 Standing tests, walking (treadmill)	10 s – 3 min. – NS	1 IMU	L5 – S1 Vert (St), right ankle (Dy)
Mancini et al. [32]	Laboratory, Home *	Daily life turning	N/A – Continuous – 7 days	3 IMUs	L5, top of left and right feet
Matsumoto et al. [47]	Laboratory	Walking self-selected pace	N/A – 9 m – NS	1 TriAx Acc	L3 Vert
Mohler et al. [34]	Home (supervised) *	2 Standing balance tests, walking Daily movement	15 s – 4.75 m – NS N/A – Continuous – 48 hours	5 IMUs	Close to Sacrum, Left and right thighs and shanks
O'Sullivan et al. [2]	Laboratory	4 Standing balance tests	30 s – N/A – NS	1 TriAx Acc	L3 Vert
Paterson et al. [48]	Laboratory *	Walking self-selected pace (circuit)	N/A – 7 min. – NS	2 TriAx Acc	Left and right 2nd metatarsals
Pozaic et al. [35]	Home	Daily life activity	N/A – Continuous – 7 days	1 TriAx Acc	Wrist
Van Schooten et al. [27]	Home *	Daily life walking	N/A – Continuous – 8 days	1 TriAx Acc	L5 Vert
Weiss et al. [49]	Laboratory	Timed-Up-and-Go test	N/A – Clinical test – 2	1 TriAx acc	L3 – L5 Vert
Weiss et al. [33]	Laboratory, Home *	Walking self-selected pace Daily life movement	N/A – 1 min – NS N/A – Continuous – 3 days	1 TriAx Acc	Lower back
High Risk					
Asai et al. [59]	Laboratory	Walking self-selected pace	N/A – 15 m – 1	2 TriAx Acc	L3 Vert, right heel
Bautmans et al. [50]	Hospital	Walking self-selected pace	N/A – 18 m – 3 × 2	1 TriAx Acc	Sacrum
Brodie et al. [51]	Laboratory	Walking self-selected pace	N/A – 21 m – 5	1 TriAx Acc	Head (helmet)
Ganea et al. [52]	Laboratory	Sit-to-stand, Stand-to-sit transition	N/A – N/A – NS	1 IMU	Sternum (skin)
Ishigaki et al. [53]	Laboratory	Free walking	N/A – 10 m – NS	1 IMU	S2 Vert
Matsumoto et al. [41]	Laboratory	Walking self-selected pace	N/A – 9 m – NS	1 TriAx Acc	L3 Vert
Menz et al. [54]	Laboratory	2 Surfaces walk self-selected pace	N/A – 20 m – 2	2 TriAx Acc	Head (helmet), sacrum
Moe-Nilssen and Helbostad, [55]	Laboratory	Fit: Walking 4 different speeds Frail: Walking 4 different speeds	Fit: N/A – 10 m – Repeatedly Frail: N/A – 6 m – Repeatedly	1 TriAx Acc	Lower back
Senden et al. [56]	Laboratory	Walking self-selected pace	N/A – 20 m – 6	1 TriAx Acc	Sacrum
Shin et al. [31]	Laboratory	2 different walking courses	N/A – N/A – 3	1 TriAx Acc	L3 Vert
Yack and Berger, [30]	Laboratory	Walking self-selected pace	N/A – 9.1 m – Repeatedly	3 UAx Acc	T2 Vert
Zakaria et al. [57]	Laboratory	Timed-Up-and-Go test	N/A – Clinical test – N/S	1 IMU	L2 Vert

Duration = Static trial duration – Dynamic trial duration/distance – Number of trials, UAx = Uniaxis, DAx = Dual axis, TriAx = Triaxial, Acc = Accelerometer, IMU = Inertial measurement unit, AP = Anteroposterior, ML = Mediolateral, L = Lumbar, C = Cervical, T = Thoracic, S = Sacrum, Vert = Vertebrae, St = Static, Dy = Dynamic, NS = Not stated, SS = self-selected, FP = Fast pace, EO = Eyes open, EC = Eyes closed, * Prospective study design.

twenty, suggesting a very low moderate risk of bias. There were 21 (60%) studies which had a moderate risk of bias and fourteen (40%) which had a low risk of bias. Zero studies were determined to have a high risk of bias (Table 6).

4. Discussion

The aim of the review was to identify characteristic differences and fall risk factors related to gait and posture, which have been measured using WIS in older adults identified as fallers and/or high fall risk. In addition to, how the measures correlated with current clinical tests, the number of sensors used, the concurrent anatomical sensor location(s) and the setting of where measurements took place, have all been identified. Through the use of a systematic methodology, the review was able to effectively explore the aforementioned aims, enabling a rigorous assessment of current literature and areas for future research.

There were 149 potential static (whilst standing still) and dynamic (whilst in motion) gait and posture differences identified that were categorised as temporal, spatial, linear acceleration, angular velocity, position and angle, and energy parameters [8]. Of the 149 differences, 127 were dynamic based measures, whilst 22 were static based measures. Wearable inertial sensor based measurements largely concurred with clinical fall risk assessments [60,61], through correlation analysis, despite being limited to five studies. Therefore, there is a clear scope for WIS to provide a good quantitative alternative to traditional fall risk observations, in the measurement of gait and postural stability for assessing fall risk among older adults [2].

4.1. Dynamic tests

Brauer et al. [62] suggests that dynamic tests provide a better prediction of falls, because most falls occur whilst an individual is in motion. Therefore, it is of no surprise that 91% of the studies within the review used a form of dynamic testing (e.g. walking). The primary

Table 3
Characteristics assessed through dynamic testing.

Category	Characteristic
Temporal Variables	Slower walking speed [25,27,29,31,33,42,43,45,49,50,51,56,57,58,59]
	Lower step/stride frequency [27,42,56,57]
	Lower stride regularity [42]
	Slower step/stride speed [28,33,49,57]
	Slower stride velocity [28]
	Less single support [28]
	Greater double support [28,34]
	Greater swing time variation [28]
	Greater stride velocity variation [28]
	Less step time variation [28]
	Slower maximum walking speed [25,59]
	Greater step time variability [29,56,59]
	Longer turn duration [32,57]
	Slower speed of turning [32]
	Lower turning cadence [57]
	Greater mean walk bout duration [34]
	Lower mean sitting bout duration [34]
	Greater Timed-Up-and-Go duration [49,57]
	Longer sit-to-stand duration [52]
	Longer stand-to-sit transition duration [52]
	Slower stand-bend time [57]
	Slower bend-sit time [57]
	Slower stand-sit time [57]
Spatial Variables	Shorter stride/step length [27,28,29,42,51,56,58]
	Lower stride symmetry [42]
	Greater number of steps/strides [43,57]
	Less steps per walk [44]
	Shorter longest walk [44]
	Greater mode variability [43]
	Greater stride length variation [28]
	Greater number of steps per turn [32,57]
	Greater CoV steps per turn [47]
	Less strides per day [27]
	Less total duration of locomotion [27]
	Lower median number of steps per bout [33]
Linear Acceleration Variables	Lower head vertical acceleration RMS [29,51]
	Lower head AP acceleration RMS [29]
	Lower head ML acceleration RMS [29]
	Greater head AP RMS [51,58]
	Greater head ML RMS [51]
	Lower pelvis vertical acceleration RMS [29]
	Lower pelvis AP acceleration RMS [29]
	Lower pelvis ML acceleration RMS [29]
	Lower lower trunk vertical acceleration RMS [47,56,57]
	Lower lower trunk AP acceleration RMS [47,57]
	Lower lower trunk ML acceleration RMS [57]
	Greater AP Amplitude (dominant hand) [35]
	Lower signal vector magnitude jerk (dominant hand) [35]
	Lower Vertical SD [27]
	Lower AP SD [27]
	Lower vertical signal range [27]
	Lower AP signal range [33]
	Lower ML signal range [33]
	Lower range sit-to-stand [49]
	Lower vertical bend-stand RMS [57]
	Lower vertical stand-bend RMS [57]
	Lower jerk sit-to-stand [49]
	Lower range stand-to-sit [49]
	Lower jerk stand-to-sit [49]
	Lower pelvic vertical acceleration [53]
	Lower pelvic AP acceleration [53]
	Lower pelvic ML acceleration [53]
	Lower inter-stride amplitude variability [56]
	Greater lower trunk ML COM acceleration (% RMS) [31]
	Lower upper trunk vertical peak acceleration [30]
	Lower upper trunk AP peak acceleration [30]
Angular Velocity Variables	Lower pelvic vertical angular velocity [53]
	Lower pelvic AP angular velocity [53]
	Lower pelvic ML angular velocity [53]
	Lower walking roll RMS angular velocity [57]
	Lower walking pitch RMS angular velocity [57]
	Lower walking yaw RMS angular velocity [57]
	Lower amplitude turning pitch angular velocity [57]
	Lower amplitude turning yaw angular velocity [57]
	Lower roll stand-bend RMS angular velocity [57]
	Lower roll bend-sit RMS angular velocity [57]
	Lower pitch stand-bend RMS angular velocity [57]
	Lower pitch bend-sit RMS angular velocity [57]
	Lower yaw stand-bend RMS angular velocity [57]
Position and Angle Variables	Lower CoV turn angle [32]
	Greater support (non-dominant hand) [35]
	Greater head 95% range AP velocity [51]
	Greater head 95% range ML velocity [51]
	Greater head AP RMS displacement [51]
	Greater head 95% range AP displacement [51]
	Lower pelvic vertical angle [53]
	Lower pelvic AP angle [53]
	Lower pelvic ML angle [53]
Energy Variables	Lower upper trunk vertical HR [30,45]
	Lower upper trunk AP HR [30,45]
	Lower lower trunk V HR [27,45,56]
	Lower lower trunk AP HR [27,45]
	Lower lower trunk ML HR [45]
	Lower mid trunk V HR [58]
	Lower mid trunk AP HR [58]
	Lower mid trunk ML HR [58]
	Greater entropy at maximum walking [25]
	Lower head V acceleration HR [29,58]
	Lower head AP acceleration HR [29]
	Lower head ML acceleration HR [29,58]
	Lower pelvis vertical acceleration HR [29]
	Lower pelvis AP acceleration HR [29]
	Lower pelvis ML acceleration HR [29]
	Greater maximum Lyapunov exponent [39]
	Lower vertical autocorrelation [47]
	Greater 90th percentile walk bout duration [34]
	Lower 90th percentile sitting bout duration [34]
	Greater ML oscillation (non-dominant hand) [35]
	Greater vertical index of harmonicity (non-dominant hand) [35]
	Lower vertical entropy (non-dominant hand) [35]
	Greater AP index of harmonicity [27]
	Lower vertical energy (non-dominant hand) [35]
	Lower ML energy (non-dominant hand) [35]
	Lower vertical amplitude of dominant frequency [33]
	Greater ML dominant frequency power [27,33]
	Greater ML slope of dominant frequency [33]
	Greater vertical logarithmic rate of divergence [27]
	Greater AP logarithmic rate of divergence [27]
	Greater ML logarithmic rate of divergence [27]
	Greater sit-to-stand local energy [52]
	Less sit-to-stand smoothness [52]
	Greater stand-to-sit local energy [52]
	Less stand-to-sit smoothness [52]
	Lower vertical step/stride autocorrelation coefficient [41,50]
	Lower AP step/stride autocorrelation coefficient [41,55]
	Greater ML step autocorrelation coefficient [55]
	Lower ML stride autocorrelation coefficient [41,50]

CoV = Coefficient of variation; RMS = Root mean square; AP = Anteroposterior; ML = Mediolateral; SD = Standard deviation; COM = Centre of mass; HR = Harmonic ratio.

dynamic gait difference between fallers and non-fallers identified through WIS coincide with observations from a clinician's fall risk assessment. During an assessment of gait, risk factors include slower walking speed and shorter step and stride lengths, which were consistently found to be characteristic differences in a number of studies. A clinical test of gait interprets fall risk based upon the duration it takes to complete a walking based task, such as the Timed-Up-and-Go test, with a longer time to completion indicating greater fall risk. It is conceivable

Table 4
Characteristics assessed through static testing.

Category	Characteristic
Linear Acceleration Variables	Greater ML RMS sway [26]
	Greater lower trunk AP RMS (EC/ECOF/ECF) [28,38]
	Greater lower trunk ML RMS (EC/ECF) [28,38]
	Greater lower trunk vertical RMS (EO/EC) [38]
	Greater ML range [37]
	Greater ankle sway (EO) [34]
	Greater hip sway (EO) [34]
	Greater COM sway (EO) [34]
	Greater ML COM sway (EO) [34]
	Greater lower trunk RMS (EO – mat) [2]
Angular Velocity Variables	Lower minimum roll angle [37]
	Greater lower trunk vertical RMS angular velocity (EO/EC) [38]
	Greater lower trunk AP RMS angular velocity (EO) [38]
	Greater lower trunk ML RMS angular velocity (EO/EC) [38]
Position and Angle Variables	Greater total displacement [37]
	Greater maximal displacement [37]
Energy Variables	Greater AP spectral edge frequency [28]
	Lower ML spectral edge frequency [28]
	Greater peak to peak amplitude [40]
	Greater 25% quartile frequency [40]
	Greater median frequency scores [40]
	Greater 75% quartile frequency [40]

RMS = Root mean square; AP = Anteroposterior; ML = Mediolateral; EO = Eyes open; EC = Eyes closed; ECF = Eyes closed foam surface.

Table 5
Number of measures obtained WIS placement.

Sensor Location	Number of Measures
Head	14
C7 Vert	1
T10 Vert	7
T2 Vert	4
Sternum	20
L3 – L5 Vert	46
L2 Vert	26
Sacrum	20
Trunk	4
Lower back	10
Iliac Crest	1
Shins	9
Ankle	1
Feet	1
Wrist	8

that a slower gait with shorter strides and steps is a conscious compensatory mechanism, as older adults adopt a more cautious gait pattern stemming from a heightened fear of falling and decreased muscular strength [28]. This is corroborated by Asai et al. [59], who reported individuals with a fear of falling walked slower than those without a fear of falling. However, as these temporal risk factors may be a conscious alteration made by older adults due to fear of falling, individuals who do not have the same fear may not exhibit such an altered gait pattern. Indeed, Mohler et al. [34] failed to identify any differences in walking speed, stride time and stride length among groups of fallers and non-fallers who were non-frail, pre-frail and frail. Nevertheless, these findings do coincide with previous non-WIS research, where slower walking speeds have been linked to increased fall risk [63]. However, a non-linear relationship between gait speed and falls has been reported [64], which suggests that a slower walking speed equates to greater fall risk when walking indoors, and when walking outdoors a faster gait speed resembles a higher risk of falls. Despite this, all studies included within the review were conducted indoors, including the seven studies conducted in the home environment. Consequently, there is a lack of information on faller and non-faller characteristics and subsequently potential fall risk outdoors. The portable nature of WIS allow for outdoor testing to provide a more complete assessment of fall risk and thus, should be incorporated into future research to establish whether fall risk factors associated with outdoor activities can be identified. Additionally, outdoor unsupervised testing will reduce the possibility of the Hawthorne effect, where subjects are consciously aware of being tested.

A consistent finding was that fallers and high fall risk older adults walked with a less smooth and less stable gait pattern, but the location of the sensor varied and there were inconsistencies as to the optimal location (upper or lower trunk). The majority of studies positioned an inertial sensor at the lower trunk, but it was the upper trunk that yielded the highest specificity for predicting future falls in a prospective study design [45]. Vertical, anteroposterior and mediolateral acceleration root mean squares (RMS) were found to be lower among fallers and high fall risk older adults, although findings were mixed on centre of mass acceleration. The acceleration signal during the walking phase is a measure of balance during gait and increased acceleration variability during steady state gait is indicative of reduced balance [65]. However, it is suggested that a lower acceleration RMS during walking represents higher gait instability [47,66,67], as individuals are walking more rigidly [57]. This coincides with Van Schooten et al. [27] who reported a higher logarithmic rate of divergence which suggests a lower local stability (greater movement from a point of equilibrium) as a prospective fall risk factor from a home based study. This perhaps confirms older adults who are fallers walk rigidly causing greater instability during gait. These WIS-based findings not only coincide with clinical observations but assist in giving clinicians further pertinent information regarding an individual's fall risk which may otherwise be missed.

4.2. Static tests

Traditionally, measurements of postural kinetics are confined to a laboratory and based upon centre of pressure movement whilst standing on a force plate [37]. This requires expensive equipment which is mounted into a surface rendering it non-portable. Nevertheless, cheaper and more commercially available alternatives such as: Nintendo Wii Balance Boards and Microsoft Kinect have both been suggested to be reliable and valid, but have mixed success regarding their capabilities [68,69]. However, despite being somewhat portable and cheaper they are still largely limited to an indoor environment and may be susceptible to greater noise and inconsistent sampling rates [69]. The most common WIS location was the posterior lower trunk between the third and fifth lumbar vertebrae (Tables 2A and 2B), and this location also yielded the most measures across all studies within this review (Table 5). Collectively the posterior lower trunk area had the most identified measures, and this coincides with Montesinos et al. [21] who also identified the lower trunk as an effective location for sensor derived features. The lower trunk, particularly the third to fifth lumbar vertebrae closely resembles an individual's centre of mass, which has been reported to correlate highly with centre of pressure platforms [26,70], suggesting WIS to be a reliable measure of balance and postural stability [71]. During static testing it was generally determined that fallers exhibit a greater amount of mediolateral and anteroposterior trunk sway than non-fallers [26,28,37,38]. Vertical motion was also identified [38] but findings were inconsistent, as naturally individuals sway forwards and backwards and side to side in an attempt to maintain balance. These findings are in agreement with current literature, as previous research without WIS have also reported similar results regarding trunk sway [72–75]. These results highlight the importance of measuring postural stability when assessing fall risk, in addition to the efficiency in which WIS can assist clinicians. This is especially so when considering that clinical tests, such as the Romberg

Table 6
Downs and Black risk of bias quality assessment.

Study	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	16	17	18	19	20	Total
Fallers																					
Alqahtani et al. [26]	1	1	1	1	0	1	1	0	0	1	1	0	0	1	1	1	1	0	0	0	**12**
Auvinet et al. [42]	1	1	1	1	0	1	1	0	0	1	1	0	0	1	1	1	1	0	0	0	**12**
Brodie et al. [43]	1	1	0	1	1	1	1	0	1	1	1	0	0	1	1	1	1	0	0	0	**13**
Brodie et al. [44]	1	1	1	1	0	1	1	0	1	1	1	0	0	0	1	1	1	0	0	0	**12**
Cho and Kamen, [40]	1	1	1	1	1	1	0	0	1	1	1	0	0	1	1	1	1	0	0	0	**13**
Cole et al. [58]	1	1	1	1	1	1	1	0	0	1	1	0	0	1	1	1	1	0	1	0	**14**
Doheny et al. [28]	1	1	0	1	0	1	1	0	1	1	1	0	0	1	1	0	1	0	0	0	**11**
Doi et al. [45]	1	1	1	1	1	1	1	0	1	1	1	0	0	1	1	1	1	0	1	0	**15**
Gago et al. [37]	1	1	1	1	1	1	1	0	1	1	1	0	0	1	1	1	1	0	1	0	**15**
Greene et al. [38]	1	0	1	1	1	1	1	0	0	1	1	0	0	1	1	1	1	0	0	0	**12**
Kojima et al. [25]	1	1	1	1	0	1	1	0	0	1	1	0	0	1	1	1	1	0	0	0	**12**
Laessoe et al. [46]	1	1	1	1	1	1	1	0	0	1	1	0	0	1	1	1	1	0	0	0	**13**
Latt et al. [29]	1	1	1	1	1	1	1	0	0	1	1	0	0	1	1	1	1	0	1	0	**15**
Liu et al. [39]	1	1	0	1	0	1	0	0	1	1	1	0	0	1	1	1	1	0	0	0	**11**
Mancini et al. [32]	1	1	1	1	1	1	1	0	1	1	1	0	0	1	1	1	1	0	1	0	**15**
Matsumoto et al. [47]	1	1	1	1	1	1	1	0	1	1	1	0	0	1	1	1	1	0	1	0	**15**
Mohler et al. [34]	1	1	1	1	1	1	1	0	1	1	1	0	0	1	1	1	1	0	1	0	**15**
O'Sullivan et al. [2]	1	1	0	1	1	1	1	0	1	1	1	0	0	1	1	1	1	0	0	0	**13**
Paterson et al. [48]	1	1	1	1	1	1	1	0	1	1	1	0	0	1	1	1	1	0	0	0	**14**
Pozaic et al. [35]	1	1	1	1	1	1	1	0	1	1	1	0	0	1	1	1	1	0	1	0	**15**
Van Schooten et al. [27]	1	1	0	1	1	1	1	0	0	1	1	0	0	1	1	1	1	0	0	0	**12**
Weiss et al. [49]	1	1	1	1	1	1	1	0	1	1	1	0	0	1	1	1	1	0	1	0	**15**
Weiss et al. [33]	1	1	1	1	1	1	1	0	1	1	1	0	0	1	1	1	1	0	1	0	**15**
High Risk																					
Asai et al. [59]	1	1	1	1	1	1	1	0	0	1	1	0	0	1	1	1	1	0	1	0	**14**
Bautmans et al. [50]	1	1	1	1	1	1	1	0	0	1	1	0	0	1	1	1	1	1	1	0	**15**
Brodie et al. [51]	0	1	1	1	1	1	1	0	1	1	1	0	0	1	1	1	1	0	0	0	**13**
Ganea et al. [52]	1	1	1	0	0	1	1	0	0	1	1	0	0	1	1	1	1	0	0	0	**11**
Ishigaki et al. [53]	1	1	1	1	0	1	1	0	0	1	1	0	0	1	1	1	1	0	0	0	**12**
Matsumoto et al. [41]	1	1	1	1	1	1	1	0	0	1	1	0	0	1	1	1	1	0	1	0	**14**
Menz et al. [54]	1	1	0	1	1	1	1	0	1	1	1	0	0	1	1	1	1	0	0	0	**13**
Moe-Nilssen and	1	1	1	1	0	1	1	0	1	1	1	0	0	1	1	1	1	0	0	0	**13**
Helbostad, [55]	1	1	1	1	0	1	1	0	1	1	1	0	0	1	1	1	1	0	0	0	**13**
Senden et al. [56]	1	1	1	1	0	1	1	0	1	1	1	0	0	1	1	1	1	0	0	0	**13**
Shin et al. [31]	1	1	0	1	0	1	1	0	0	1	1	0	0	1	1	1	1	0	0	0	**11**
Yack and Berger, [30]	1	1	0	1	0	1	1	0	0	1	1	0	0	1	1	1	1	0	0	0	**11**
Zakaria et al. [57]	1	1	0	1	0	1	1	0	0	1	1	0	0	1	1	1	1	0	0	0	**11**

Questions in brief: (1) Aim clearly described, (2) Outcomes clearly described, (3) Subjects clearly described, (4) Measurement clearly described, (5) Distribution of confounders clearly described, (6) Main findings clearly described, (7) Estimates of random variability, (8) Important adverse events reports, (9) P value reported, (10) Subjects asked represent the population, (11) Subjects used represent the population, (12) Were examiners blinded, (13) Data dredging, (14) Appropriate statistical tests used, (15) Valid and reliable outcome measures, (16) Subjects from same population, (17) Subjects recruited from same time period, (18) Measures in a random order, (19) Adjustments for confounding variables, (20) Sufficient power.
Marked out of 20. 1 = yes and 0 = no. High risk of bias = 0–6, moderate risk of bias = 7–13, low risk of bias = 14–20.

test, rely on observing postural sway under different standing conditions [76].

4.3. WIS based measures compared to clinical testing

There were five studies that presented correlation analysis on WIS derived measures to clinical fall risk assessments with varying but largely promising results. Low and weak correlations were observed for maximum entropy [25] and mediolateral harmonic ratios [43] to walking speed. However, there were strong statistically significant correlations between the TUG and Berg Balance Scale tests to the RMS of the acceleration signal and angular velocity signals during bouts of quite standing [2,38]. Additionally, and perhaps even more promisingly, Senden et al. [56] reported all acceleration based measures correlated with the Tinetti scale. This included strong positive correlations for walking speed and step length, two commonly identified gait differences among fallers and non-fallers which have been suggested as fall risk factors. These strong positive correlations provide further scope of the efficacy and ability of WIS to perform and be applied under clinical fall risk assessments. However, despite providing initial promising results, only three of five studies provided strong correlations between WIS measures and clinical fall risk assessments. Therefore, future studies should incorporate correlation analysis to provide further strength to this notion. Interestingly, the Timed-Up-and-Go test (a commonly used clinical assessment) was not able to distinguish between fallers and non-fallers when measured with a stopwatch, but was when using acceleration derived measures [49]. This finding is consistent within the literature as stopwatch derived Timed-Up-and-Go test duration has previously been identified to not always be a sensitive marker of fall risk among older adults [77,78]. These results suggest WIS are a better tool when assessing fall risk, even when stopwatch durations are similar, coupled with their low-cost portable nature make WIS a desirable option for fall risk assessment in a clinical environment, as they also do not require the reactions of an assessor to start and stop a test.

There were six studies which included a prospective study design. These studies suggest gait and postural based differences to be fall risk factors as opposed to just characteristic differences between fallers and non-fallers. One prospective study in particular, Mancini et al. [32], identified that the quality of turning measured as the coefficient of variation for the number of steps per turn, could distinguish and predict future fallers and non-fallers over a six month period, even when both the Tinetti gait and Tinetti balance tests could not. This suggests that WIS capabilities have the potential to go above and beyond that of

current clinical testing. Although, the overall results of the review and the individual studies themselves hold great meaning, more prospective studies are needed within the literature. This is because, many gait and postural based differences between fallers and non-fallers could stem from a fear of falling or even a gait adaption resulting from a fall itself. Therefore, identifying pre-existing fall risk factors will be pertinent to the eventual prevention of future falls.

There was no study included within the review deemed to have a high risk of bias. Information reporting and external validity scored generally well. However, the internal validity and selection bias of studies were not as strong. This could be a consequence of questions regarding assessors being blind to testing and randomisation of measures being difficult to perform with WIS. Additionally, no study reported sample size power calculations, potentially reducing the statistical power of available data.

A limitation to the present review, is that only measures that were identified to be statistically significant in at least one paper were highlighted as a gait and posture difference among fallers and non-fallers, even if that measure was not statistically significant in another paper. Although, it must be noted that this rarely occurred, and more often than not, measures were only included within a single study. Whilst the present study being a systematic scoping review rather than a meta-analysis, could be viewed as a limitation, the aim was to undertake a descriptive synthesis. In addition, the heterogeneity of the included studies precluded the possibility of meta-analyses.

5. Conclusion

The review undertook a comprehensive search of the literature and was able to demonstrate that WIS have the capabilities to objectively identify fall risk among older adults, even when clinical tests could not. It was determined that spatiotemporal measures such as: slower walking speeds, shorter step and stride lengths, in addition to, acceleration based metrics including reduced acceleration RMS were most related to fall risk during dynamic tasks. Whereas, increased trunk sway, often measured with the RMS of the acceleration signal was most commonly related to fall risk during static tasks. However, it should be noted that most studies were of a retrospective study design and as such, highlight characteristic differences between fallers and non-fallers which may not necessarily infer fall risk factors. Consequently, more prospective studies are needed. Greater RMS of the acceleration signal appears to be strongly related with faster walking speeds, however, no study directly reported this. The review also included studies (6 total) where a cohort may have had a pathological disorder (e.g., Parkinson's and Alzheimer's), and findings would suggest the same metrics may relate to fall risk regardless of this. This suggests WIS can effectively be used among different cohorts including older healthy and those with neurological ailments. Nevertheless, as such a small sample of pathological disorder studies were identified, it is difficult to make a definitive conclusion and future studies should explore this. A single WIS located at the lower trunk was frequently witnessed and was highly successful in determining several gait and posture differences among older adults. There was a lack of studies outside of a laboratory and during daily life. Therefore, it is currently unknown what the most accurate methods measuring gait metrics are in this new setting. However, regarding WIS anatomical locations it appears to be the same as in laboratory testing, and this also has a very high compliance rate even when multiple WIS are used. Whilst measures of gait during daily life appear to be similar to testing in a laboratory, there was scope for additional metrics to be observed during daily life, such as turning during gait, which provided pertinent information regarding prospective falls even when current clinical assessments could not. A comparison of metrics of gait during continuous daily life monitoring and outdoors, to metrics of gait inside a laboratory was not identified by any of the studies who included a continuous daily life monitoring aspect. Consequently, how these metrics compare in relation to fall risk is currently unknown. Therefore, future studies should consider how metrics of gait compare indoors and outdoors, which would provide greater insight regarding fall risk among older adults. It was determined that WIS are an effective tool to be implemented into current clinical fall risk assessments and perhaps more so in continuous unsupervised habitual monitoring. The lightweight portable nature enables WIS to be used in outdoor testing, which at present, has been largely neglected. This will provide a greater wealth of information regarding fall risk, which is currently unknown. The use of WIS will enable clinicians and healthcare practitioners to make more informed objective decisions regarding fall risk, and its application should be considered in future fall risk assessments.

Ethics approval and right to participate

N/A.

Consent for publication

All authors provide consent for publication.

Availability of data and material

This research was a scoping review with a systematic methodology, and therefore, no data is available.

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Author contributions

MP conceived the idea for the review which was developed along with VG and AP. MP and VG developed a search strategy. MP undertook database searches and reviewed all articles along with VG, who acted as second reviewer. MP wrote the initial and final drafts of the manuscript. MP, VG and AP altered the draft of the manuscript to form the final version. All authors read and approved the final manuscript.

Declaration of Competing Interests

There are no competing interests.

References

[1] N.K. Sekaran, H. Choi, R.A. Hayward, K.M. LangaFall, Associated difficulty with activities of daily living (ADL) in functionally independent older adults aged 65 to 69 in the United States: a cohort study, J. Am. Geriatr. Soc. 61 (2013) 96–100.
[2] M. O'Sullivan, C. Blake, C. Cunningham, C. Finucane, G. Boyle, Correlation of accelerometry with clinical balance tests in older fallers and non-fallers, Age Ageing 38 (2009) 308–313.
[3] B. Resnick, E.S. Nahm, S. Zhu, C. Brown, M. An, B. Park, J. Brown, The impact of osteoporosis, falls, fear of falling and efficacy expectations on exercise among community dwelling older adults, Orthop. Nurs. 33 (2014) 277–286.
[4] M.E. Tinetti, M. Speechley, S.F. Ginter, Risk factors for falls among elderly persons living in the community, N. Engl. J. Med. 319 (1988) 1701–1707.
[5] P. Pasquetti, L. Apicella, G. Mangone, Pathogenesis and treatment of falls in elderly, Clin. Cases Miner. Bone Metab. 11 (2014) 222–225.
[6] Y. Dionyssiotis, Analyzing the problem of falls among older people, Int. J. Gen. Med. 5 (2012) 805–813.
[7] A.C. Scheffer, M.J. Schuurmans, N. van Dijk, T. van der Hooft, S.E. de Rooij, Fear of falling: measurement strategy, prevalence, risk factors and consequences among older persons, Age Ageing 37 (2008) 19–24.
[8] J. Howcroft, J. Kofman, E.D. Lemaire, Review of fall risk assessment in geriatric

populations using inertial sensors, J. Neuroeng. Rehabil. 10 (2013) 91.
[9] R.Y.W. Lee, A.J. Carlisle, Detection of falls using accelerometers and mobile phone technology, Age Ageing 40 (2011) 690–696.
[10] E.R. Burns, J.A. Stevens, R. Lee, The direct costs of fatal and non-fatal falls among older adults - United States, J. Safety Res. 58 (2016) 99–103.
[11] D. Podsiadlo, S. Richardson, The timed "Up & Go": a test of basic functional mobility for frail elderly persons, J. Am. Geriatr. Soc. 39 (1991) 142–148.
[12] M. Csuka, D.J. McCarty, Simple method for measurement of lower extremity muscle strength, Am. J. Med. 78 (1985) 77–81.
[13] A. Shumway-Cook, M.H. Woollacott, Motor Control: Theory and Practical Applications, Williams & Wilkins, 1995.
[14] K. Berg, S. Wood-Dauphine, J.I. Williams, D. Gayton, Measuring balance in the elderly: preliminary development of an instrument, Physiother. Canada 41 (1989) 304–311.
[15] M.E. Tinetti, Performance-oriented assessment of mobility problems in elderly patients, J. Am. Geriatr. Soc. 34 (1986) 119–126.
[16] M. Mancini, L. King, A. Salarian, L. Holmstrom, J. McNames, F.B. Horak, Mobility lab to assess balance and gait with synchronized body-worn sensors, J. Bioeng. Biomed. Sci. (Suppl. 1) (2011) 007.
[17] E. Giannouli, O. Bock, S. Mellone, W. Zijlstra, Mobility in old age: capacity is not performance, Biomed Res. Int. 2016 (2016) 8.
[18] R. McCarney, J. Warner, S. Iliffe, R. van Haselen, M. Griffin, P. Fisher, The Hawthorne effect: a randomised, controlled trial, BMC Med. Res. Methodol. 7 (2007) 30.
[19] L.E. Cofré Lizama, M. Pijnappels, G.H. Faber, P.N. Reeves, S.M. Verschueren, J.H. van Dieën, Age effects on mediolateral balance control, PLoS One 9 (2014) e110757.
[20] J.K. Lee, S.N. Robinovitch, E.J. Park, Inertial sensing-based pre-impact detection of falls involving near-fall scenarios, Ieee Trans. Neural Syst. Rehabil. Eng. 23 (2015) 258–266.
[21] L. Montesinos, R. Castaldo, L. Pecchia, Wearable inertial sensors for fall risk assessment and prediction in older adults: a systematic review and meta-analysis, IEEE Trans. Neural Syst. Rehabil. Eng. 26 (2018) 573–582.
[22] V.A. Goodwin, R.A. Abbott, R. Whear, A. Bethel, O.C. Ukoumunne, J. Thompson-Coon, K. Stein, Multiple component interventions for preventing falls and fall-related injuries among older people: systematic review and meta-analysis, BMC Geriatr. 14 (2014) 15.
[23] H. Arksey, L. O'Malley, Scoping studies: towards a methodological framework, Int. J. Soc. Res. Methodol. 8 (2005) 19–32.
[24] S.H. Downs, N. Black, The feasibility of creating a checklist for the assessment of the methodological quality both of randomised and non-randomised studies of health care interventions, J. Epidemiol. Commun. Health 52 (1998) 377–384.
[25] M. Kojima, S. Obuchi, O. Henmi, N. Ikeda, Comparison of smoothness during gait between community dwelling elderly fallers and non-fallers using power spectrum entropy of acceleration time-series, J. Phys. Ther. Sci. 20 (2008) 243–248.
[26] B.A. Alqahtani, M.A. Ferchak, T.J. Huppert, E. Sedjic, S. Perera, S.L. Greenspan, P.J. Sparto, Standing balance and strength measurements in older adults living in residential care communities, Aging Clin. Exp. Res. 29 (2016) 1021–1030.
[27] K.S. van Schooten, M. Pijnappels, S.M. Rispens, P.J.M. Elders, P. Lips, J.H. van Dieen, Ambulatory fall-risk assessment: amount and quality of daily-life gait predict falls in older adults, J. Gerontol. A Biol. Sci. Med. Sci. 70 (2015) 608–615.
[28] E.P. Doheny, B.R. Greene, T. Foran, C. Cunningham, C.W. Fan, R.A. Kenny, Diurnal variations in the outcomes of instrumented gait and quiet standing balance assessments and their association with falls history, Physiol. Meas. 33 (2012) 361–373.
[29] M.D. Latt, H.B. Menz, V.S. Fung, S.R. Lord, Acceleration patterns of the head and pelvis during gait in older people with Parkinson's disease: a comparison of fallers and nonfallers, J. Gerontol. A Biol. Sci. Med. Sci. 64 (2009) 700–706.
[30] H.J. Yack, R.C. Berger, Dynamic stability in the elderly: identifying a possible measure, J. Gerontol. 48 (1993) 225–230.
[31] S.S. Shin, D.H. An, W.G. Yoo, Effects of balance recovery after square and semi-circular turns on gait velocity and center of mass acceleration in older adults with differing visual acuity, Eur. Geriatr. Med. 6 (2015) 103–108.
[32] M. Mancini, H. Schlueter, M. El-Gohary, N. Mattek, C. Duncan, J. Kaye, F.B. Horak, Continuous monitoring of turning mobility and its association to falls and cognitive function: a pilot study, J. Gerontol. A Biol. Sci. Med. Sci. 71 (2016) 1102–1108.
[33] A. Weiss, M. Brozgol, M. Dorfman, T. Herman, S. Shema, N. Giladi, J.M. Hausdorff, Does the evaluation of gait quality during daily life provide insight into fall risk? A novel approach using 3-day accelerometer recordings, Neurorehabil. Neural Repair 27 (2013) 742–752.
[34] M.J. Mohler, C.S. Wendel, R.E. Taylor-Piliae, N. Toosizadeh, B. Najafi, Motor performance and physical activity as predictors of prospective falls in community-dwelling older adults by frailty level: application of wearable technology, Gerontology 62 (2016) 654–664.
[35] T. Pozaic, A.K. Grebe, U. Lindemann, W. Stork, Sit-to-Stand transition reveals acute fall risk in activities of daily living, IEEE J. Transl. Eng. Health Med. 4 (2016) 1–11.
[36] M.A.D. Brodie, M.J.M. Coppens, S.R. Lord, N.H. Lovell, Y.J. Gschwind, R.J. Redmond, M.B. Del Rosario, K. Wang, D.L. Sturnieks, M. Persiani, K. Delbaere, Wearable pendant device monitoring using new wavelet-based methods shows daily life and laboratory gaits are different, Med. Biol. Eng. Comput. 54 (2016) 663–674.
[37] M.F. Gago, V. Fernandes, J. Ferreira, H. Silva, L. Rocha, E. Bicho, N. Suso, Postural stability analysis with inertial measurement units in alzheimer's disease, Dement. Geriatr. Cogn. Dis. Extra 4 (2014) 22–30.
[38] B.R. Greene, D. McGrath, L. Walsh, E.P. Doheny, D. McKeown, C. Garattini, C. Cunningham, L. Crosby, B. Caufield, R.A. Kenny, Quantitative falls risk estimation through multi-sensor assessment of standing balance, Physiol. Meas. 33 (2012) 2049–2063.
[39] J. Liu, X. Zhang, T.E. Lockhart, Fall risk assessments based on postural and dynamic stability using inertial measurement unit, Saf. Health Work 3 (2012) 192–198.
[40] C. Cho, G. Kamen, Detecting balance deficits in frequent fallers using clinical and quantitative evaluation tools, J. Am. Geriatr. Soc. 46 (1998) 426–430.
[41] H. Matsumoto, H. Hagino, M. Osaki, S. Tanishima, C. Tanimura, A. Matsuura, T. Makabe, Gait variability analysed using an accelerometer is associated with locomotive syndrome among the general elderly population: the GAINA study, J. Orthop. Sci. 21 (2016) 354–360.
[42] B. Auvinet, G. Berrut, C. Touzard, L. Moutel, N. Collet, D. Chaleil, E. Barrey, Gait abnormalities in elderly fallers, J. Aging Phys. Act. 11 (2003) 40–52.
[43] M.A.D. Brodie, K. Delbaere, S.R. Lord, H.B. Menz, S.T. Smith, Good lateral harmonic stability combined with adequate gait speed is required for low fall risk in older people, Gerontology 61 (2014) 69–78.
[44] M.A. Brodie, S.R. Lord, M.J. Coppens, J. Annegarn, K. Delbaere, Eight-week remote monitoring using a freely worn device reveals unstable gait patterns in older fallers, IEEE Trans. Biomed. Eng. 62 (2015) 2588–2594.
[45] T. Doi, S. Hirata, R. Ono, K. Tsutsumimoto, S. Misu, H. Ando, The harmonic ratio of trunk acceleration predicts falling among older people: results of a 1-year prospective study, J. Neuroeng. Rehabil. 10 (2013) 7.
[46] U. Laessoe, H.C. Hoeck, O. Simonsen, T. Sinkjaer, M. Voigt, Fall risk in an active elderly population–can it be assessed? J. Negat. Results Biomed. 6 (2007) 2.
[47] H. Matsumoto, T. Makabe, T. Morita, K. Ikuhara, A. Kajigase, Y. Okamoto, E. Ashikawa, E. Kobayashi, H. Hagnno, Accelerometry-based gait analysis predicts falls among patients with a recent fracture who are ambulatory: a 1-year prospective study, Int. J. Rehabil. Res. 38 (2015) 131–136.
[48] K. Paterson, K. Hill, N. Lythgo, Stride dynamics, gait variability and prospective falls risk in active community dwelling older women, Gait Posture 33 (2011) 251–255.
[49] A. Weiss, T. Herman, M. Plotnik, M. Brozgol, N. Giladi, J.M. Hausdorff, An instrumented timed up and go: the added value of an accelerometer for identifying fall risk in idiopathic fallers, Physiol. Meas. 32 (2011) 2003–2018.
[50] I. Bautmans, B. Van Keymolen, T. Mets, B. Jansen, Reliability and clinical correlates of 3D-accelerometry based gait analysis outcomes according to age and fall-risk, Gait Posture 33 (2011) 366–372.
[51] M.A. Brodie, C.G. Canning, T.R. Beijer, S.R. Lord, Uncontrolled head oscillations in people with Parkinson's disease may reflect an inability to respond to perturbations while walking, Physiol. Meas. 36 (2015) 873–881.
[52] R. Ganea, A. Paraschiv-Ionescu, C. Bula, S. Rochat, K. Aminian, Multi-parametric evaluation of sit-to-stand and stand-to-sit transitions in elderly people, Med. Eng. Phys. 33 (2011) 1086–1093.
[53] N. Ishigaki, T. Kimura, Y. Usui, K. Aoki, N. Narita, M. Shimizu, K. Nara, N. Ogihara, K. Nakamura, H. Kato, M. Ohira, Y. Yokokawa, K. Miyoshi, N. Murakami, S. Okada, Y. Nakamura, N. Saito, Analysis of pelvic movement in the elderly during walking using a posture monitoring system equipped with a triaxial accelerometer and a gyroscope, J. Biomech. 44 (2011) 1788–1792.
[54] H.B. Menz, S.R. Lord, R.C. Fitzpatrick, Acceleration patterns of the head and pelvis when walking are associated with risk of falling in community-dwelling older people, J. Gerontol. A Biol. Sci. Med. Sci. 58 (2003) M446–452.
[55] R. Moe-Nilssen, J.L. Helbostad, Interstride trunk acceleration variability but not step width variability can differentiate between fit and frail older adults, Gait Posture 21 (2005) 164–170.
[56] R. Senden, H.H. Savelberg, B. Grimm, I.C. Heyligers, K. Meijer, Accelerometry-based gait analysis, an additional objective approach to screen subjects at risk for falling, Gait Posture 36 (2012) 296–300.
[57] N.A. Zakaria, Y. Kuwae, T. Tamura, K. Minato, S. Kanaya, Quantitative analysis of fall risk using TUG test, Comput. Methods Biomech. Biomed. Engin. 18 (2015) 426–437.
[58] M.H. Cole, M. Sweeney, Z.J. Conway, T. Blackmore, P.A. Silburn, Imposed faster and slower walking speeds influence gait stability differently in parkinson fallers, Arch. Phys. Med. Rehabil. 98 (2017) 639–648.
[59] T. Asai, S. Misu, R. Sawa, T. Doi, M. Yamada, The association between fear of falling and smoothness of lower trunk oscillation in gait varies according to gait speed in community-dwelling older adults, J. Neuroeng. Rehabil. 14 (2017) 5.
[60] B. Najafi, K. Aminian, F. Loew, Y. Blanc, P.A. Robert, Measurement of stand-sit and sit-stand transitions using a miniature gyroscope and its application in fall risk evaluation in the elderly, IEEE Trans. Biomed. Eng. 49 (2002) 843–851.
[61] M.R. Narayanan, B.G. Celler, S.J. Redmond, N.H. Lovell, M.E. Scalzi, S.R. Lord, Longitudinal falls-risk estimation using triaxial accelerometry, IEEE Trans. Biomed. Eng. 57 (2010) 534–541.
[62] S.G. Brauer, Y.R. Burns, P. Galley, A prospective study of laboratory and clinical measures of postural stability to predict community-dwelling fallers, J. Gerontol. A Biol. Sci. Med. Sci. 55 (2000) M469–M476.
[63] D.A. Kegelmeyer, A.D. Kloos, K.M. Thomas, S.K. Kostyk, Reliability and validity of the tinetti mobility test for individuals with parkinson disease, Phys. Ther. 87 (2007) 1369–1378.
[64] L. Quach, A.M. Galica, R.N. Jones, E. Procter-Gray, B. Manor, M.T. Hannan, L.A. Lipsitz, The non-linear relationship between gait speed and falls: the MOBILIZE boston study, J. Am. Geriatr. Soc. 59 (2011) 1069–1073.
[65] M. Henriksen, H. Lund, R. Moe-Nilssen, H. Bliddal, B. Danneskiod-Samsoe, Test-retest reliability of trunk accelerometric gait analysis, Gait Posture 19 (2004) 288–297.
[66] S. Nishiguchi, M. Yamada, K. Nagai, S. Mori, Y. Kajiwara, T. Sonoda, K. Yoshimura, H. Yoshitomi, H. Ito, K. Okamoto, T. Ito, S. Muto, T. Ishihara, T. Aoyama, Reliability and validity of gait analysis by android-based smartphone Telemedicine, J. E Health

18 (2012) 292–296.
[67] M. Yamada, T. Aoyama, S. Mori, S. Nishiguchi, K. Okamoto, T. Ito, S. Muto, H. Yoshitomi, H. Ito, Objective assessment of abnormal gait in patients with rheumatoid arthritis using a smartphone, Rheumatol. Int. 32 (2012) 3869–3874.
[68] R.A. Clark, B.F. Mentiplay, E. Hough, Y.H. Pua, Three-dimensional cameras and skeleton pose tracking for physical function assessment: a review of uses, validity, current developments and kinect alternatives, Gait Posture 68 (2019) 193–200.
[69] R.A. Clark, B.F. Mentiplay, Y.H. Pua, K.J. Bower, Reliability and validity of the Wii Balance Board for assessment of standing balance: a systematic review, Gait Posture 61 (2018) 40–54.
[70] S.L. Whitney, J.L. Roche, G.F. Marchetti, C.C. Lin, D.P. Steed, G.R. Furman, M.C. Musolino, M.S. Redfern, A comparison of accelerometry and center of pressure measures during computerized dynamic posturography: a measure of balance, Gait Posture 33 (2011) 594–599.
[71] R. Moe-Nilssen, J.L. Helbostad, Trunk accelerometry as a measure of balance control during quiet standing, Gait Posture 16 (2002) 60–68.
[72] B.E. Maki, P.J. Holliday, A.K. Topper, A prospective study of postural balance and risk of falling in an ambulatory and independent elderly population, J. Gerontol. 49 (1994) M72–M84.
[73] M. Piirtola, P. Era, Force platform measurements as predictors of falls among older people - a review, Gerontology 52 (2006) 1–16.
[74] M. Leandri, S. Cammisuli, S. Cammarata, L. Baratto, J. Campbell, M. Simonini, M. Tabaton, Balance features in Alzheimer's disease and amnestic mild cognitive impairment, J. Alzheimer Dis. 16 (2009) 113–120.
[75] A. Merlo, D. Zemp, E. Zanda, S. Rocchi, F. Meroni, M. Tettamanti, A. Recchia, U. Lucca, P. Quadri, Postural stability and history of falls in cognitively able older adults: the Canton Ticino study, Gait Posture 36 (2012) 662–666.
[76] A. Khasnis, R. Gokula, Romberg's test, J. Postgrad. Med. 49 (2003) 169–172.
[77] D. Shahar, M. Levi, I. Kurtz, S. Shany, I. Zvili, E. Mualleme, A. Shahar, O. Sarid, I. Melzer, Nutritional status in relation to balance and falls in the elderly: a preliminary look at serum folate, Ann. Nutr. Metab. 54 (2009) 59–66.
[78] G. Thrane, R.M. Joakimsen, E. Thornquist, The association between timed up and go test and history of falls: the Tromso study, BMC Geriatr. 7 (2007) 1.